After the targeted volume **tt** is well imaged, as illustrated in FIG. 7A, the method then can further define a certain margin **m** surrounding the tumor that is targeted for the ablative treatment. The working end **122** is introduced to the desired position as depicted in FIG. 7A. With the engagement plane **125** in contact with the targeted tissue, (at time T_0), the operator actuates a switch **155** that delivers Rf energy from the radiofrequency generator or source **150A** to the core conductive element or electrode **140C**. At ambient tissue temperature, the low base resistance of the medial conductive matrix **140B** allows unimpeded Rf current flow from the source **150A** through the engagement surface **125** and tissue to return electrical lead **158** that is coupled to ground pad **160**. In FIG. 7A, it can be understood that the engaged tissue **tt** that is in contact with the engagement surface **125** initially will have a substantially uniform impedance (indicated at particular resistance level **Ω**) to electrical current flow, which resistance **Ω** could increase substantially in proximity to the engagement surface **125** of the contacted tissue is overly dehydrated by the active Rf delivery.

After the initial activation of energy delivery at time T_0 as depicted in FIG. 7A, the Rf current will create a certain energy density (or active Rf energy application) in the targeted tissue. Following an arbitrary interval indicated at time T_1 in FIG. 7B, the tissue's impedance proximate to engagement surface **125** typically will be elevated to a somewhat higher impedance level due to dehydration. However, at time T_1 in FIG. 7B, the active Rf energy application that elevates the tissue temperature will instantly conduct heat to the working end **122**, including the PTC conductive layer **140B**. Thus, it can easily be understood that when the tissue temperature and the temperature of the medial PTC conductive layer **140B** reaches the level of the switching range (i.e., 68° C. to 72° C.), the Rf current flow from the core conductive electrode **140C** to the engagement surface **140A** will be substantially reduced or terminated due to the exponential increase in the resistance of medial conductor material **140B** (see FIG. 6). It is believed that such an instant and automatic reduction of Rf energy application will prevent any substantial dehydration of tissue proximate to the engagement plane **125**. By thus maintaining the desired level of moisture around the engagement plane **125**, the working end can more effectively apply energy to the tissue—and provide a deeper thermal effect than would be possible with prior art Rf needles that can cause an irreversible dehydration (impedance increase) about the working end.

Still referring to FIG. 7B, as the tissue temperature proximate to engagement surface **125** falls by thermal relaxation in the tissue and lack of an Rf energy density, the temperature of the medial conductor **140B** will thus fall below the threshold of the selected switching range. This effect then will cause Rf current to again flow through the assembly of conductive layers **140C**, **140B** and **140A** to the targeted tissue to again increase the tissue temperature by active Rf heating of the tissue. The thermal relaxation in the tissue can be highly variable and is most greatly affected by blood flow, which subtracts heat from the tissue. In hypervascularized tumor tissue, such thermal relaxation is increased in speed.

By the above described mechanisms of causing the medial conductive matrix **140B** to hover about its selected switching range, the actual Rf energy density in the tissue **tt** thus can be precisely modulated to maintain the desired temperature. FIG. 7B illustrates exemplary isotherms that can be maintained over any selected period of time to ablate the tumor and the desired tissue margins **m**. Of particular interest, the polymer matrix that comprises the medial conductor portion **140B** is doped with materials to resistively heat the matrix as Rf energy flow therethrough is reduced. Thus, the thermal mass of the working end **122** is elevated in temperature to thereby deliver energy to the targeted tissue **tt** by means of greater *passive* conductive heating—at the same time Rf energy delivery causes lesser tissue heating. This balance of active Rf heating and *passive* conductive (or radiative) heating can maintain the targeted temperature for any selected time interval.

In summary, one method of the invention comprises the delivery of Rf energy from an Rf source **150A** to a conductive engagement surface portion **140A** of a probe through a thermally sensitive resistor material (medial layer **140B**) wherein the resistor material has a selected switching range that approximates a targeted temperature of the therapy. In operation, the working end *automatically* modulates active Rf energy density in the tissue as the temperature of the engaged tissue conducts heat back to the thermally sensitive resistor material **140B** to cause its temperature to reach the selected switching range. In this range, the Rf current flow will be reduced, with the result being that the tissue temperature can be maintained in the selected range without the need for thermocouples or any other form of feedback circuitry mechanisms to modulate Rf power from the source. Most important, it is believed that this method of the

invention will allow for more immediate modulation of *actual* energy application to tissue than provided by a temperature sensor. Such temperature sensors suffer from a time lag. Further, a temperature sensor provides only an indirect reading of actual tissue temperature—since a typical sensor can only measure the temperature of the electrode.

Another method of the invention comprises providing the working end with a suitable cross-section of thermally resistive matrix **140B** so that when it is elevated in temperature to the switching range, the conductive matrix **140B** effectively functions as a resistive electrode to passively conduct thermal energy to engaged tissue. Thus, in operation, the working end **122** can automatically modulate the application of energy to tissue between *active* Rf heating and *passive* conductive heating of the targeted tissue at a targeted temperature level.

FIG. 7C illustrates another aspect of the method of the invention that relates to the Rf source **150A** and controller **150B**. A typical commercially available radiofrequency generator has feedback circuitry mechanisms that control power levels depending on the feedback of impedance levels of the engaged tissue. FIG. 7C is a graph relating to the probe of present invention that shows: (i) the temperature-resistance profile of the targeted tissue, (ii) the resistance-resistance profile of the PTC conductive matrix **140B** of the probe, and (iii) the combined resistance-resistance profile of the tissue **tt** and the PTC conductive matrix. As can be understood from FIG. 7C, in operation, the Rf source **150A** and controller **150B** can read the combined impedance of the tissue **tt** and the PTC conductive layer which will thus allow the use of the instrument with any typical Rf source without interference with feedback circuitry components.

3. Type "B" probe for energy delivery to targeted tissue. An exemplary Type "B" probe **200** corresponding to the invention is illustrated in FIG. 8 that is adapted for energy delivery to tissue and again is described in treating a targeted benign or malignant tumor. The probe **200** includes a proximal handle (not shown) coupled to an introducer portion **210** that can carries at least one extendable energy delivery member. In the exemplary embodiment of FIG. 8, the probe **200** carries a plurality of energy delivery members **220A-220B** which can number from two to 8 or more. For convenience, the probe of FIG. 8 depicts two members **220A-220B** that define respective engagement planes **225A-225B**.